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A STUDY OF AUSCULTATORY BLOOD PRESSURES IN SIMULATED ARTERIES

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SUMMARY

After considerable preliminary analysis of the pulsatile viscous flow in flexible tubes, a large-scale pulsating flow system has been constructed and various simulated arteries of 1-inch inside diameter have been tested under varying conditions of pulse rate, pressure level, and pulse pressure using, first, various mixtures of water and glycerol and, finally, whole steer blood as the fluid. The oscillatory intra-arterial pressures were recorded by pressure transducers which were attached to both static and total pressure probes in the flow itself, and these are compared with measurements of the systolic and diastolic pressures which were obtained by the clinical auscultatory technique, using a stethoscope and a specially designed pressure chamber. In addition, simultaneous measurements were made of arterial wall displacements and the velocity of the pulse wave along the tube. The simulated system has exhibited many of the qualitative features of a living human subject, and the entire spectrum of characteristic Korotkoff sounds was produced in the artery.

The test results indicate that the systolic pressure as determined with a sphygmomanometer is consistently higher than the peak intra-arterial pressure by an amount which depends on the stiffness and thickness of the arterial wall. The auscultatory diastolic pressure reading, however, may be either higher or lower than the minimum intra-arterial pressure, depending primarily on the pulse rate. This result is independent of whether the fourth or fifth phase of the Korotkoff sounds is used as an

indication of diastolic pressure. The auscultatory readings with whole steer blood are in agreement with those using water and glycerol for the same conditions. Measured arterial wall displacements and wave velocities are in approximate agreement with theory except during transient changes in mean pressure level.

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The work to be discussed here has been carried on at Vidya, a Division of Itek Corporation, in Palo Alto, California; and it has been jointly sponsored, first by the U. S. Air Force and more recently by NASA. The work is concerned with an investigation of the precise meaning (from an engineering standpoint) of the so-called systolic and diastolic blood pressures ordinarily measured by a physician using the auscultatory technique. In practice, these blood pressure measurements are made by applying a pneumatic cuff to the patient's upper arm and inflating the cuff by means of a hand pump until there is no pulse in the radial artery at the wrist. The physician then applies a stethoscope to the brachial artery just below the cuff and starts to decompress the cuff slowly. As he decompresses, at some pressure level, he begins to hear a faint tapping sound (first phase of the Korotkoff sounds) and at this point he records the pressure as the systolic pressure reading. As the cuff pressure is reduced further, the sound changes in character (second phase), becomes clearer and

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increases in intensity (third phase), then fades rather suddenly (fourth phase), and finally disappears altogether (fifth phase). The pressure corresponding to the cessation of sounds is customarily recorded by the physician as the diastolic pressure reading. Figure 1 shows a qualitative plot of the cuff pressure and the Korotkoff sound intensity (as heard in the stethoscope) as functions of time during the decompression of the cuff. Starting at the left, at zero time, is the occluding pressure, at which point there is no flow through the tube and hence, no sound is produced. Now, according to Erlanger [1], at some point below the occluding pressure, a part of the fluid passes through the tube on each cycle and impacts on the downstream stationary fluid, thereby producing a water hammer and a consequent sound. In addition, however, the sound must radiate in the radial direction and is attenuated at a rate that depends on the viscosity of the fluid. The sound is then picked up by the arterial wall, which responds at its natural frequency, and it is a combination of these sounds which is heard in the stethoscope. Beyond the first phase, the sound then increases in intensity and, as shown on the lower diagram of the figure, it reaches a peak intensity after which a sudden drop-off occurs (the fourth phase), at which point some physicians take the diastolic reading. At a still lower pressure, the sounds disappear entirely, going below the threshold of audibility, and this is the fifth phase, which is usually taken as the diastolic pressure reading. (This point will be discussed later.) Since the arterial wall responds at its natural frequency, it seems clear that if the natural frequency of the arterial wall is below the audible level of about 20 cycles per second, it will be difficult to hear the Korotkoff sounds. This is typical of young children and of patients in hypothermia.

The purpose of the present investigation is to determine the relationship (if any) between the blood pressure readings obtained by the auscultatory technique and the actual oscillatory intra-arterial pressures that existed in the vessel <u>before</u> the cuff was applied. According to the medical literature, the systolic pressure reading corresponds to the peak of this oscillatory pressure, and the diastolic pressure corresponds to the minimum of this oscillatory pressure.

Now, in order to understand the problem at hand, we must first attempt to determine the cuff pressure required to stop the oscillatory flow in the artery. As a first step, then, let us consider the pressure required to occlude the flexible tube with no flow in it at all. According to the theory of elastic stability [2] the critical pressure, p_{cr}, required (in excess of the internal pressure in the tube) for initial buckling of the tube is given by

$$p_{cr} = \frac{E}{4(1 - \sigma^2)} \left(\frac{h}{R}\right)^3$$

where E is the elastic modulus of the tube, h/R is the wall thickness-to-radius ratio, and σ is Poisson's ratio for the tube material. However, a higher pressure is clearly required to bring the diammetrically opposite sides of the tube together. The analysis of Halphen [3] and more recently of McIvor [4] indicate that this pressure is given by

$$p = 1.75\ p_{cr}$$

and a still higher pressure, for which there is no theory at present, is required to collapse the tube entirely. The cross sections of the flexible tube at various applied external pressures are illustrated in Figure 2. For complete collapse, as shown at the extreme right, the classical theoretical method for solving such buckling problems fails because of the nonlinear character of the problem associated with large deformations and appreciable bending moments. Even the solution of this problem, however, would at best yield only the occluding pressure for a tube with no oscillatory flow in it. The theoretical determination of the auscultatory systolic and diastolic pressures is considerably more complicated, and no theoretical analysis has yet been attempted.

An experimental approach is pursued here by use of a simulated system in which control of each of the parameters can be accomplished independently. Of course, in resorting to a simulated system, one must observe the rules of proper modeling or scaling of the system in order to obtain meaningful results. In other words, there should be dynamic similarity between the simulated system and the actual physiological system. For the present problem, this involves matching the Reynolds number based on mean flow, the Reynolds number based on frequency and the radius of the tube, the ratio of the mean pressure level to the elastic modulus of the tube, the ratio of the pulse pressure or pressure amplitude to the mean pressure, the wall thickness-to-radius ratio, the natural frequency of the tube, and Poisson's ratio for the tube material. In addition, if the material is viscoelastic, as we know natural arteries to be, then one should also match the dynamic elastic modulus compared with the static elastic modulus. Since most of the above parameters are not accurately known for the human brachial artery, the emphasis in the present investigation is on the variation of each of the non-dimensional parameters over a wide range covering the physiological range of interest insofar as it is known. For the idealized model studied here, gum rubber tubing is used to represent the artery, and the fluid initially is represented by a combination of water and glycerol at varying concentrations in order to vary the viscosity of the fluid and thereby the Reynolds numbers of the flow. Experiments using real blood will be discussed later.

The experimental apparatus is shown schematically in Figure 3. The apparatus is a closed loop system in which the test fluid is fed from the reservoir through the circulation pump to the pulse generator section. With the pulse generator operating, periodic pressure pulses are imparted to the test liquid which then divides into two branches. These two branches represent (1) the test artery and (2) that part of the human system that is not closed off when the oscillatory flow in the artery is cut off by application of the pressure cuff. In order to avoid mutual interference of the pressure waves in the two branches, the test section and the bypass have separate venous return lines that connect back to the reservoir. The gum rubber "artery" is 10 feet long and has a nominal inside diameter of 1 inch and a wall thickness of 1/16 inch.

After the experimental apparatus had been assembled, the first futile attempts to produce the Korotkoff sounds by using a physician's pneumatic cuff led to a re-examination of Erlanger's work done in 1916 [1]. From Erlanger's work, it became apparent that a large-volume pressure chamber, properly scaled up to our system, was required. The volume of air in the chamber must be above a certain minimum value relative to the size of the artery, and the diaphragms used in the chamber to compress the artery must be sufficiently flexible to permit a pressure pulse to pass through the artery. In addition, the pressure chamber must not impose any artificial constraints on the artery at the ends of the chamber.

An exploded view of the resulting pressure chamber is shown in Figure 4, where only the parts of the lower section of the chamber are identified since the upper and lower sections are identical. The top and bottom parts of the pressure chamber are essentially open hardwood boxes with outside dimensions $18 \times 12 \times 2$ inches. A rubber "o" ring gasket is imbedded in the wall of the box to provide a seal, and a thin rubber diaphragm covers the opening of each box. A plastic frame 1 inch thick is placed over each diaphragm as shown in the figure. The plastic frames have semi-circular openings in their opposite ends to permit the artery to pass freely through the pressure chamber so that the chamber does not exert any artificial "end effects" on the artery.

Using this pressure chamber, our consulting physician, Dr. David Bruns*, who is a cardiovascular specialist, was able to hear the entire spectrum of Korotkoff sounds and obtain the associated systolic and diastolic pressure readings. It was noted that an auscultatory gap could be produced at either very high pulse rates and/or high mean pressure levels, and changes in quality and intensity of the Korotkoff sounds ordinarily associated with such clinical conditions as hypertension and aortic insufficiency could be produced by changes in the flow conditions or in the arterial wall. Now, since the auscultatory readings do depend to some degree upon the auditory acuity of the individual observer, each of the observers for the present experiments was trained by Dr. Bruns to recognize and distinguish the sounds at the onset of the Korotkoff sounds (systolic pressure) and at the fourth and fifth phases (diastolic pressure). All readings were made by at least two observers, and agreement within 5 mm Hg was obtained in all cases.

Since the present investigation is made to compare the auscultatory pressure readings with direct pressure readings inside

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the artery, the instrumentation for determining the actual oscillatory pressure inside the tube becomes important. Figure 5 shows a pressure traverse mechanism which consists of a pressure probe mounted in such a way that it can be moved radially inside the test artery, by means of a micrometer adjustment, to any vertical location relative to the fixed table. The frame on which the probe is mounted is the same frame that clamps the pressure chamber into place (see Fig. 7) and insures that both measurements are made at the same axial location in the tube. The measurements made with this traverse mechanism indicated that there was no appreciable variation of pressure with radial position through the tube. Thus, a fixed probe installation was designed and used in all the tests. But, in order to make the measurements required to compare the auscultatory readings with direct readings, the pressure probes must be calibrated in their proper invironment. For the present problem this is of no minor concern, since at very low Reynolds numbers an impact tube may not read the actual impact pressure, because of viscous dissipation ahead of the stagnation point. In addition, we have an oscillatory flow in which the probe may be subject to a time lag. And finally, over part of the cycle the probe will be operating in reverse flow, so that it will read the pressure in its own wake, which may be less than free-stream pressure. Therefore, in order to calibrate for all these effects, a special calibration rig (Fig. 6) was designed and constructed. A similar rig was originally used for steady flow by Hurd, Cheske, and Shapiro [5]. But for the present experiments, it has been adapted for unsteady effects and for reverse flow over the probe. This device is simply a rotating arm on which the probe to be used is mounted and is immersed in an annular tank containing the test fluid. The arm can be rotated at variable speeds and accelerations by means of a falling weight. The instantaneous speed of rotation is recorded by monitoring the shadow of the spoke falling on a photocell, and from this record the velocities and accelerations can be calculated. The pressure reading from the probe is recorded simultaneously and is compared with the known pressure head given by the depth of the probe below the surface of liquid. These probe calibration tests for both viscous and acceleration effects indicated a maximum error of 2 percent for total pressure and 1 percent for static pressure over the range of the present tests. Hence, these errors were neglected and the measured probe readings were taken as correct within the accuracy with which the blood pressure readings could be made.

In the present tests, two pressure probes were mounted downstream from (i.e., distal to) the pressure chamber, as shown in Figure 7. One probe measures static (lateral) pressure and the other total (stagnation) pressure inside the tube as a function of time. These fixed probes were installed approximately on the axis of the test artery by inserting them from the bottom of the tube in order not to disturb the tube during pulsation. The pressure outputs from the probes are recorded on an oscillograph by the use of battery-powered transducers. A check of the pressure readings obtained with these probes against readings obtained at the location of the pressure chamber showed that the error in pressure level and pulse amplitude was less than 2 percent. There will, of course, also be a phase shift due to the downstream location of the probes, but this is of no concern for the present investigation.

The variable-stroke piston arrangement shown in Figure 8 is provided with an air chamber which separates the actual working fluid from the driving mechanism. The piston itself is mounted on a rolling diaphragm for quiet operation. The pressure in the chamber is adjusted to give a fine control on the volumetric displacement and the pulse pressure amplitude. The main purpose of this chamber is to provide a means of obtaining a large pulse pressure amplitude with a small volumetric displacement.

The actual experiments that were carried out can perhaps best be illustrated by a brief description of the test procedure. The reservoir height is first adjusted to give the required mean pressure level in the artery. Then the stroke of the piston is adjusted to obtain the desired pulse pressure amplitude, and the pulse generator apparatus is adjusted for motor speed to obtain the desired pulse rate. The pressure in the air chamber, which separates the actual test fluid from the pulse pump, is adjusted to give a finer control on the volumetric displacement and the pulse pressure amplitude and wave shape. It should be noted that all the experimental data presented here are for a sinusoidal pulse wave.

Analytical expressions relating the pulse wave velocity to the oscillatory pressures and wall displacements in flexible tubes have been derived using the theoretical work of Womersley [6] and of Morgan and Kiely [7]. Therefore, in order to check these relationships, the pulse wave velocity was measured in the test artery by using two photocells to follow the wall displacement of the tube at two axial locations as functions of time. (The wave velocity is determined from the phase difference between the two photocell outputs.)

The auscultatory pressure measurements are made by trained engineers by pressurizing the pressure cuff with the gas bottle, applying the stethoscope snugly to the simulated artery just distal to the pressure cuff, and recording the pressures observed on the manometer when the first and fifth phases of the Korotkoff sounds occur. The oscillograph record consists of two photocell outputs (giving wall displacements), two pressure traces (for the intra-arterial pressures), and one timing pulse for frequency determinations. The viscosity of the test fluid, and the radius of the artery as a function of the mean internal pressure, are also determined just prior to the actual tests. From these inputs, the static and total pressures inside the artery are obtained as functions of time, the wave velocity is calculated, and finally the unsteady Reynolds number $R^2(\omega/\nu)$ is determined.

The results of our experiments are shown, in part, in Figure 9 for a 1-inch diameter, gum rubber simulated artery of 1/16 inch wall thickness. This figure is a plot of the systolic pressure readings (obtained by the auscultatory technique) divided by the maximum oscillatory static (lateral) pressure in the tube. This ratio is plotted as a function of the parameter $R\sqrt{\omega/\nu}$ where R is the mean tube radius, ω the pulse frequency, and ν the fluid kinematic viscosity. This parameter was shown in the analyses of Womersley [6] and of Morgan and Kiely [7] to be a fundamental parameter; and it is seen that the ratio plotted in Figure 9 is relatively independent of the Reynolds number based on frequency, but is consistently higher than unity, meaning that the systolic pressure reading is consistently higher than the actual maximum static pressure in the tube. This is in accordance with our earlier statements regarding the buckling pressure required to collapse the tube itself. This point is illustrated further by the fact that we made tests with a thicker wall tube (1/8 inch wall), as indicated by the square points on the plot, and here obtained still higher systolic readings for the same flow conditions, as expected.

The corresponding results for the diastolic pressure are shown in Figure 10. Here we have the ratio of the diastolic pressure reading to the minimum oscillatory static pressure. Unlike the systolic results, we do not see a consistent trend with the Reynolds number based on frequency. In fact, the ratio here is seen to be highly frequency-dependent; that is, if the patient has a very low pulse rate, he may indicate a very high diastolic pressure reading. On the other hand, if he has a high pulse rate, he may indicate a very low diastolic pressure reading for the same flow conditions. (The approximate range of unsteady Reynolds number for the human brachial artery is indicated by the shaded region of the plot.) Because of this somewhat peculiar behavior with unsteady Reynolds number, we investigated the possibility that we should have taken the diastolic reading at the fourth phase rather than at the fifth phase of the Korotkoff sounds. Some runs were repeated, taking the fourth phase as the diastolic reading, and one of these points is indicated by the half-shaded circle. Other repeat runs (at various mean pressures and frequencies) indicated that the situation was not essentially improved, and there seems to be no fundamental difference in the frequency dependence whether the fourth or fifth phase is taken except, of course, that the fourth phase gives consistently higher readings than the fifth.

An obvious question in regard to the foregoing results is whether or not they are applicable to real blood. It is recalled that the test fluid was a combination of water and glycerol to provide a wide range of viscosity. In order to check the generality of these results, we considered the possibility of modeling the red blood cells by means of plastic or glass beads, but decided that these were not a satisfactory representation of the red cells for this problem, since the diffusion, reflection, and absorption of sound would be quite different for such solid particles than for the actual flexible red cells. Therefore, we resorted to the use of real blood. However, since our system requires 250 gallons of fluid, we were forced to confine the blood to the test section itself, and this was done by installing toy balloons at either end of the test section. On one side of the balloon is the blood and on the other side (and throughout the rest of the system) is water. Before this system was used in actual test runs, it was tested both with the same fluid and with two different fluids on the two sides, to insure that the presence of the balloons did not affect the auscultatory readings. We then re-ran specific cases using fresh steer blood in the simulated artery. For these tests, we had the assistance of two medical technicians from the Palo Alto-Stanford Hospital. Fresh steer blood was obtained each morning. Then, at the end of each day, the system was sterilized in a special ultra-violet light chamber, after being thoroughly scrubbed with Haemosol and rinsed. Hematocrit readings and sedimentation rates were taken for the blood used, and we found essentially a zero sedimentation rate which is typical of steer blood, and which simplified the testing, since no stirring was required. The hematocrit averaged about 45 percent, and the specific gravity was 1.06.

Results for the real blood obtained in this manner are shown by the solid symbols in the plots of systolic and diastolic pressure ratios against unsteady Reynolds number (see Figs. 9 and 10). It is seen that the results for real blood are in essential agreement with those obtained using combinations of water and glycerol.

The conclusions from this experimental investigation are that (1) the systolic and diastolic pressure readings as obtained by the auscultatory technique are not, in general, the same as the actual systolic and diastolic pressures obtained by direct intra-arterial pressure probes, (2) the systolic pressure reading obtained by the auscultatory technique is consistently higher than the peak pressure in the tube, because of the pressure required to collapse the tube itself. The degree to which it is too high depends on the stiffness and thickness of the arterial wall, and does not appear to be frequency-dependent, (3) the auscultatory diastolic pressure reading may be either too high or too low, depending strongly on the pulse rate (this conclusion is independent of whether the fourth or fifth phase is used for the diastolic pressure reading), and (4) the results using real steer blood are in essential agreement with those obtained using water and glycerol as the blood fluid. Finally, it should be mentioned that our velocity and wall displacement measurements indicate fairly good agreement with the theoretical results of Womersley and of Morgan and Kiely, except for the transient case in which the mean pressure level is changing with time.

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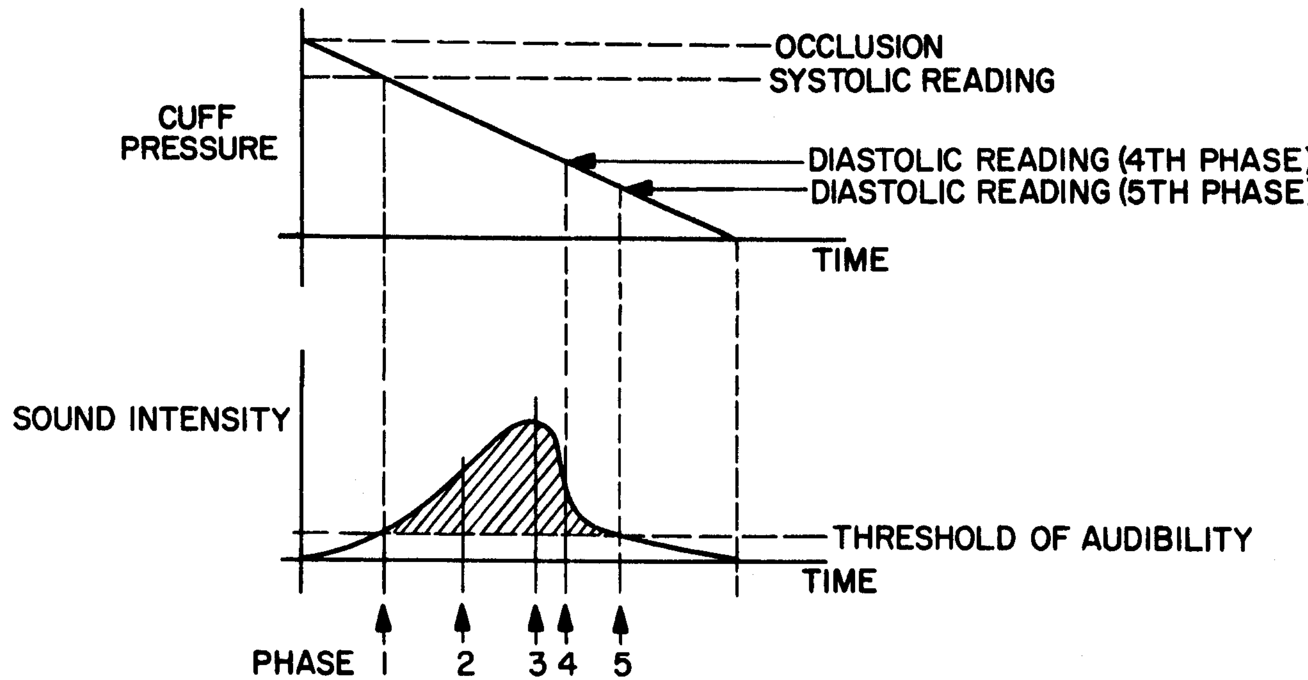

Figure 1.- Variation of intensity of Korotkoff sounds during cuff decompression.

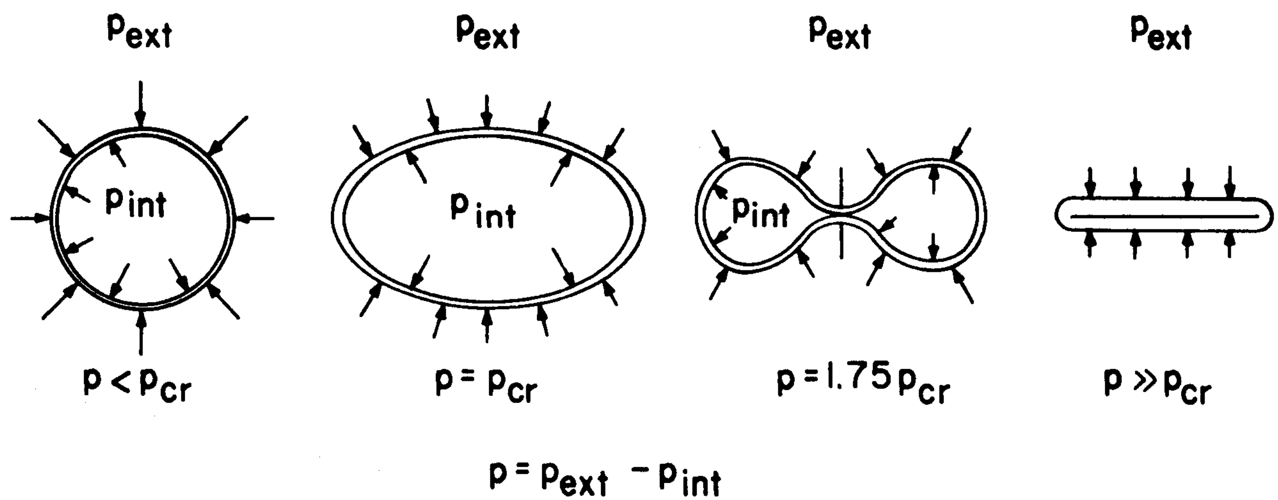

Figure 2.- Collapse of thin wall tube due to external pressure.

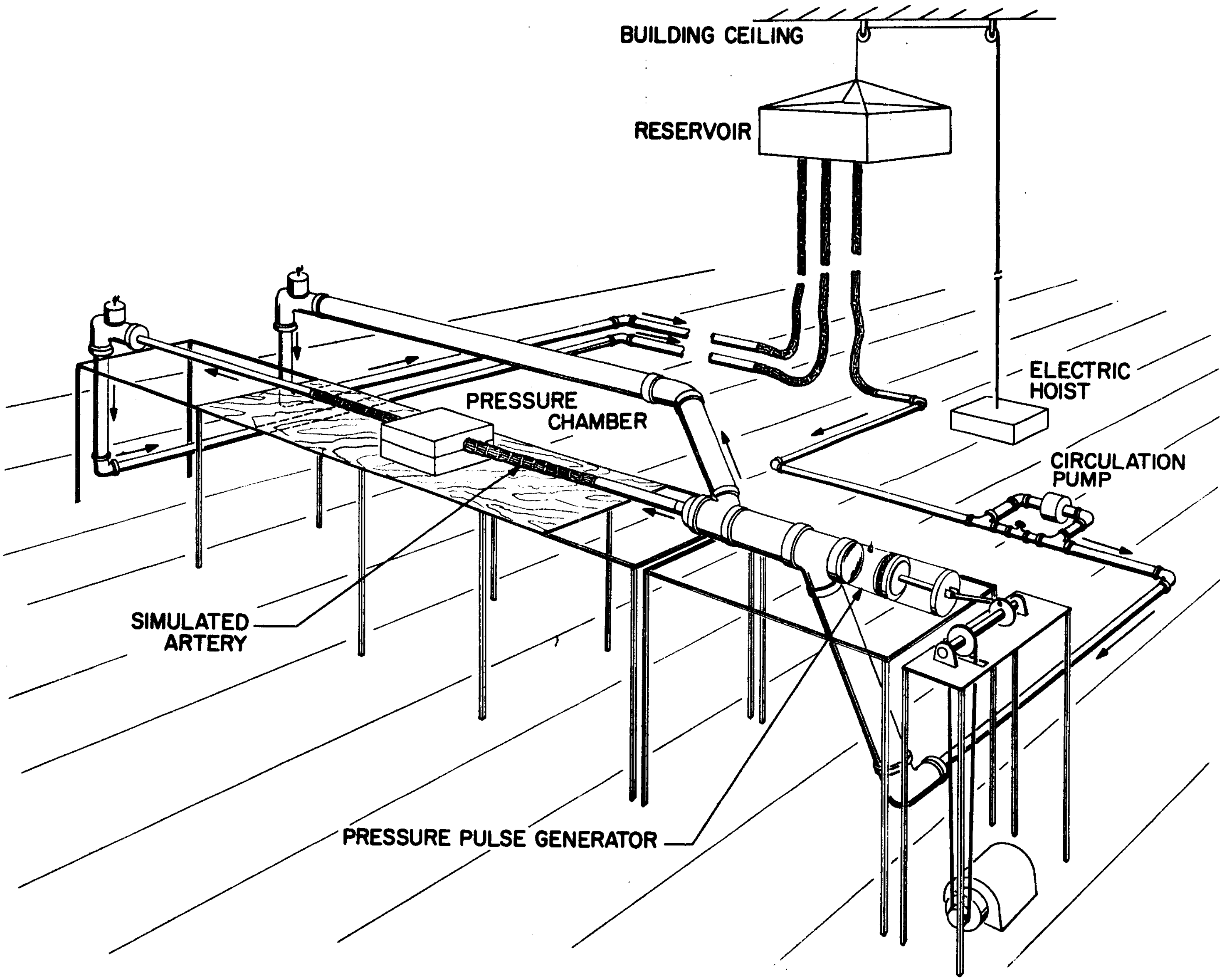

Figure 3.- Schematic sketch of experimental apparatus.

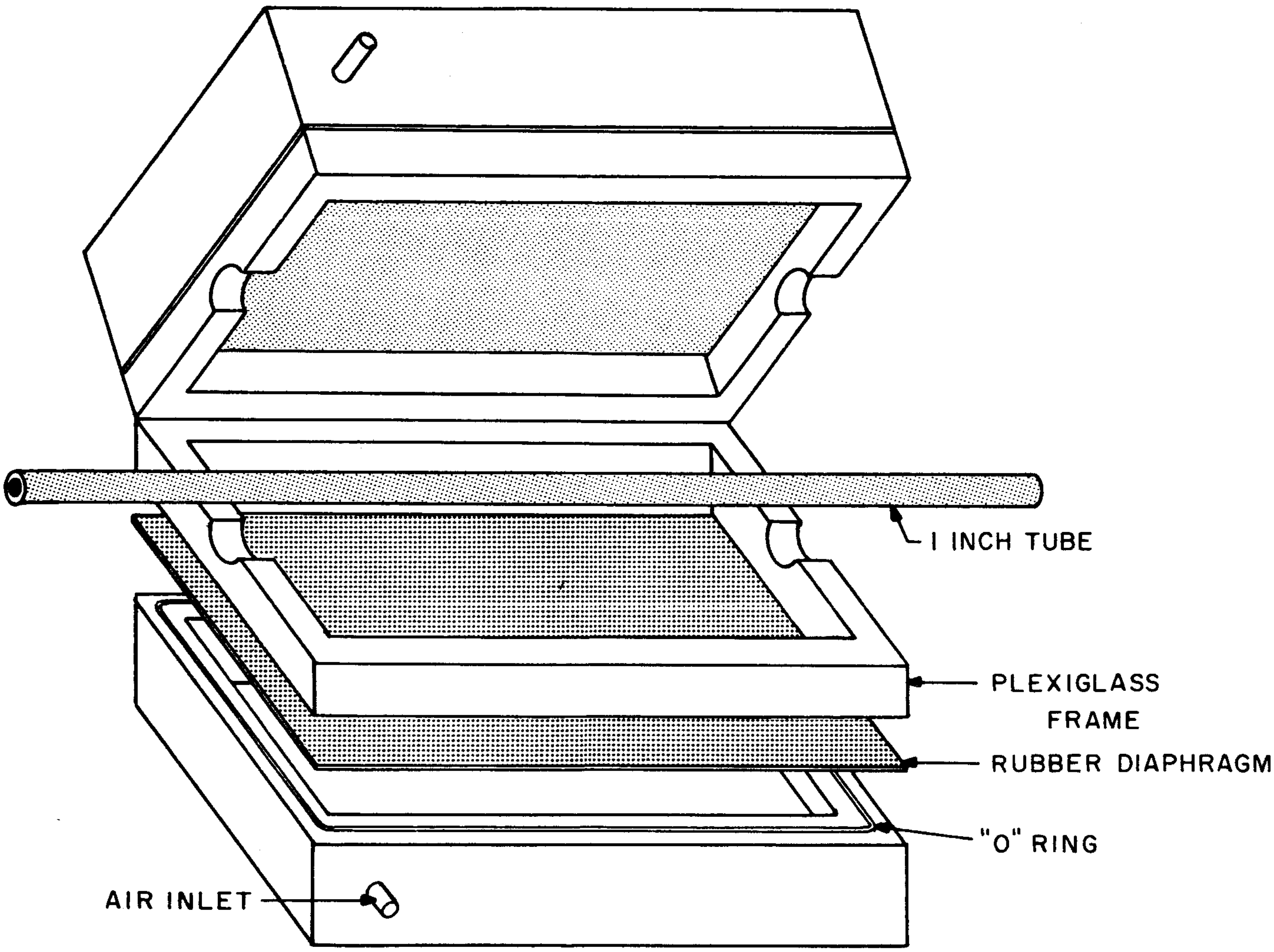

Figure 4.- Exploded view of pressure chamber used in present tests.

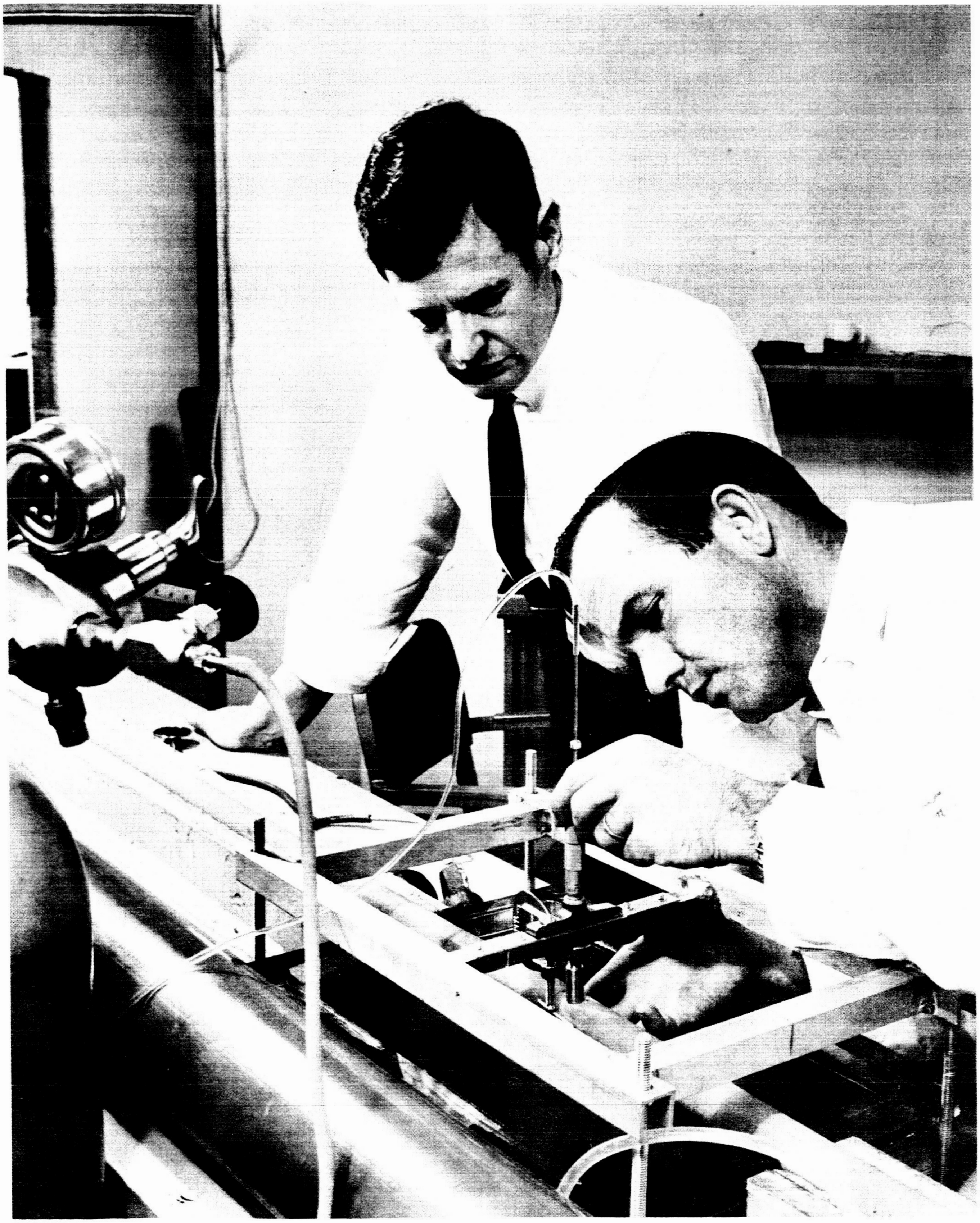

Figure 5.- Installation of pressure probe traverse mechanism on pressure chamber mounting.

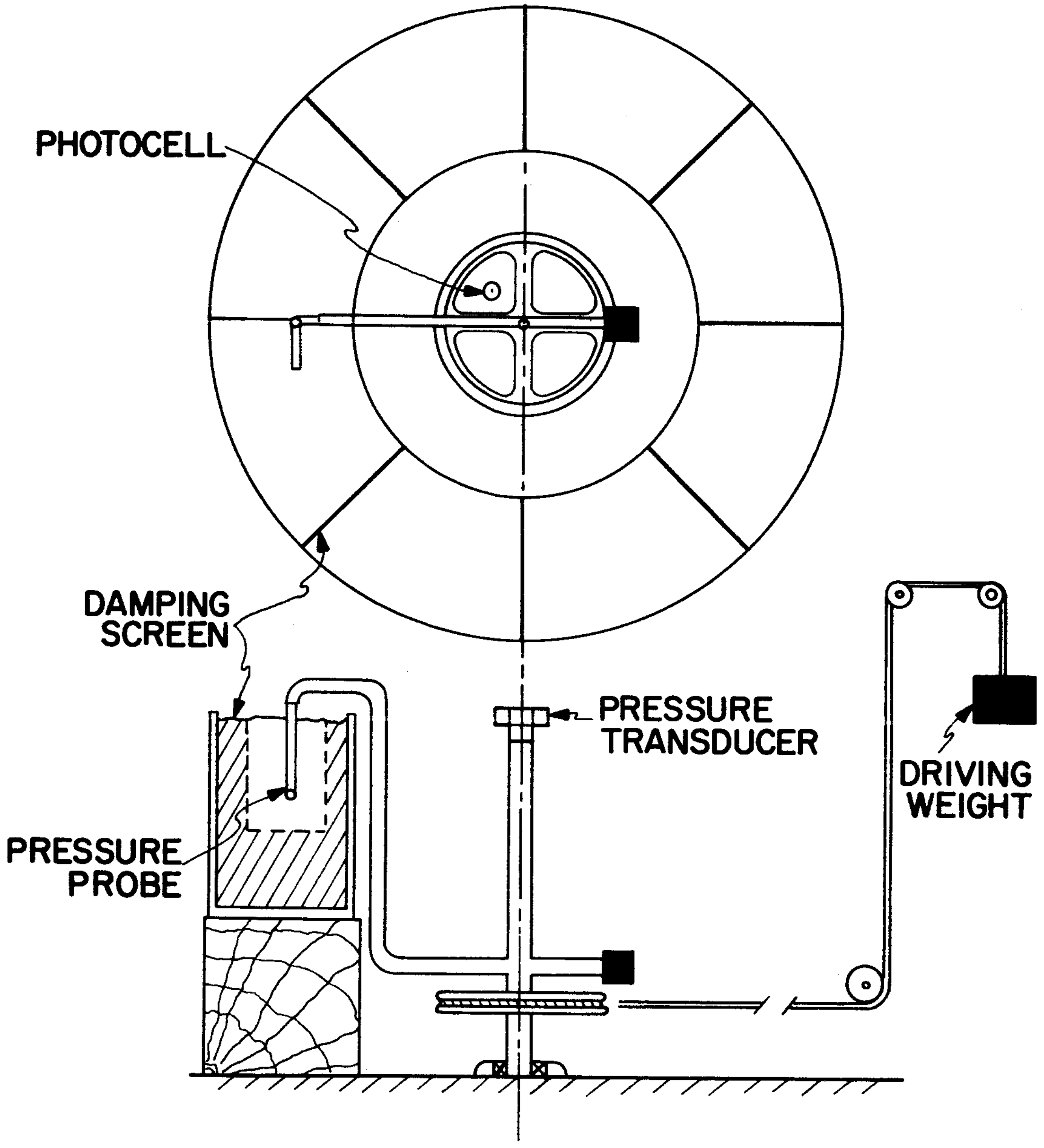

Figure 6.- Calibration rig for pressure probes.

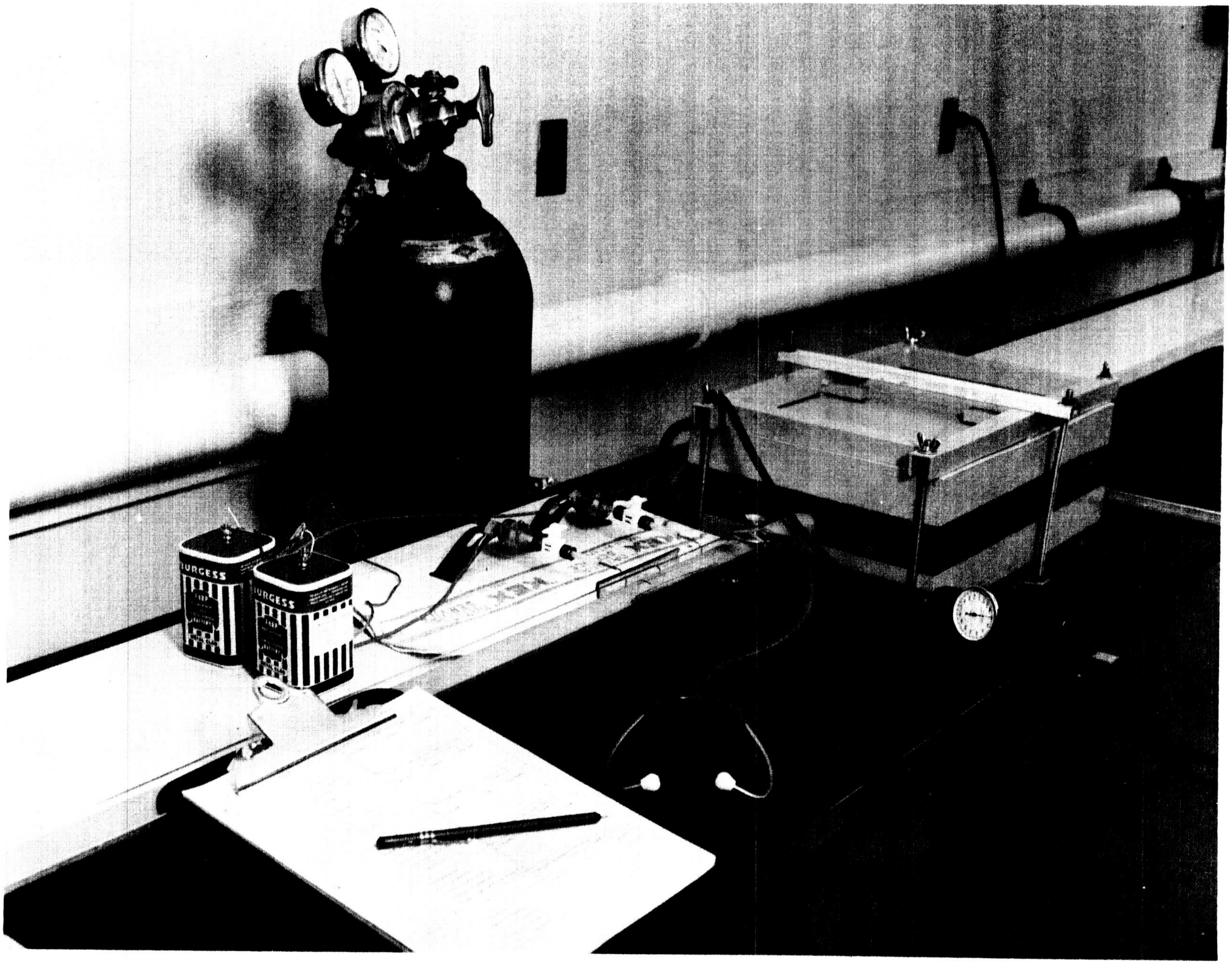

Figure 7.- Equipment used for auscultatory and direct blood pressure measurements on simulated artery.

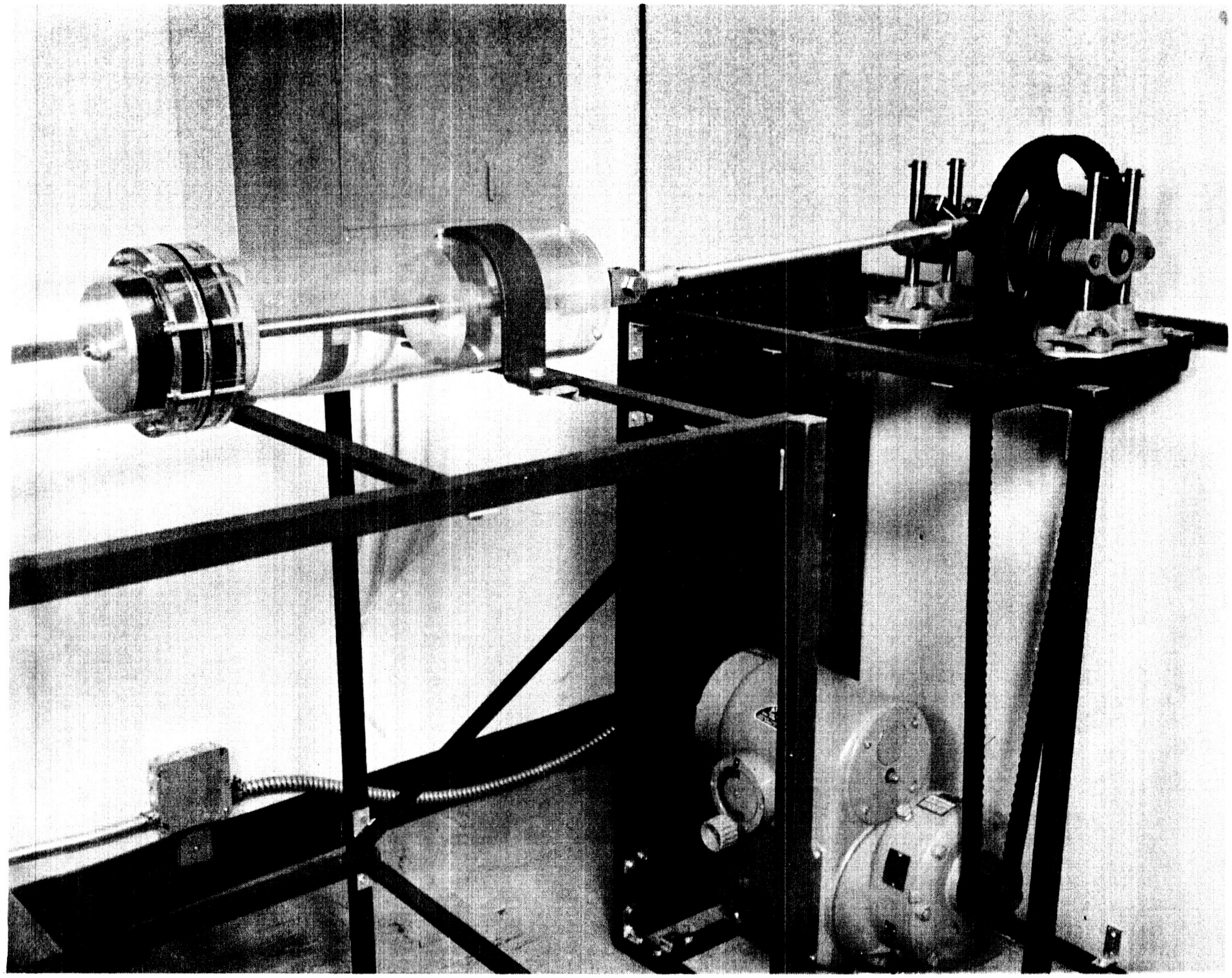

Figure 8.- Variable amplitude, variable frequency sinusoidal pulse generator.

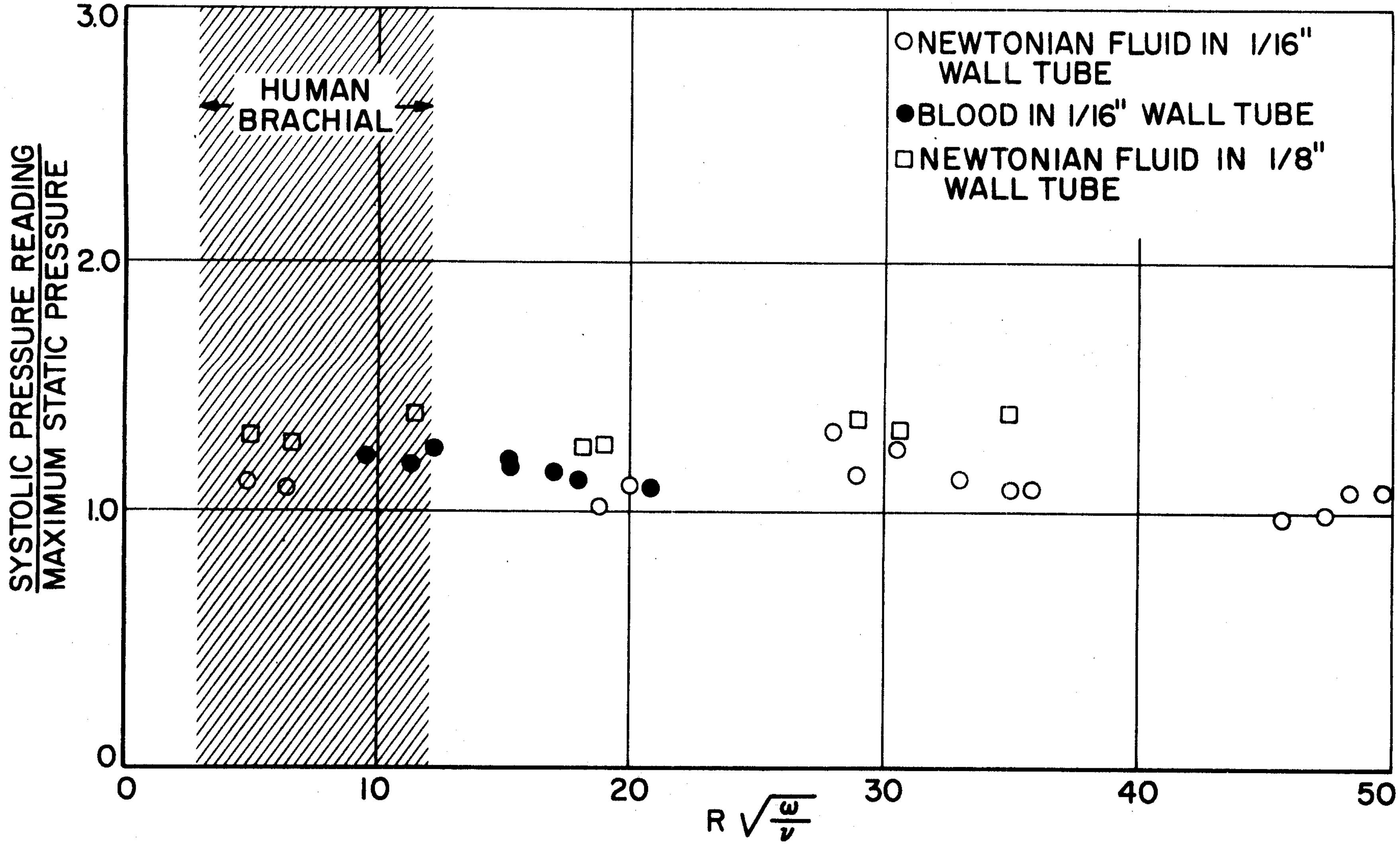

Figure 9.- Variation of systolic pressure readings with frequency parameter for simulated arteries (mean pressure 155 mm Hg, pulse pressure 48 mm Hg).

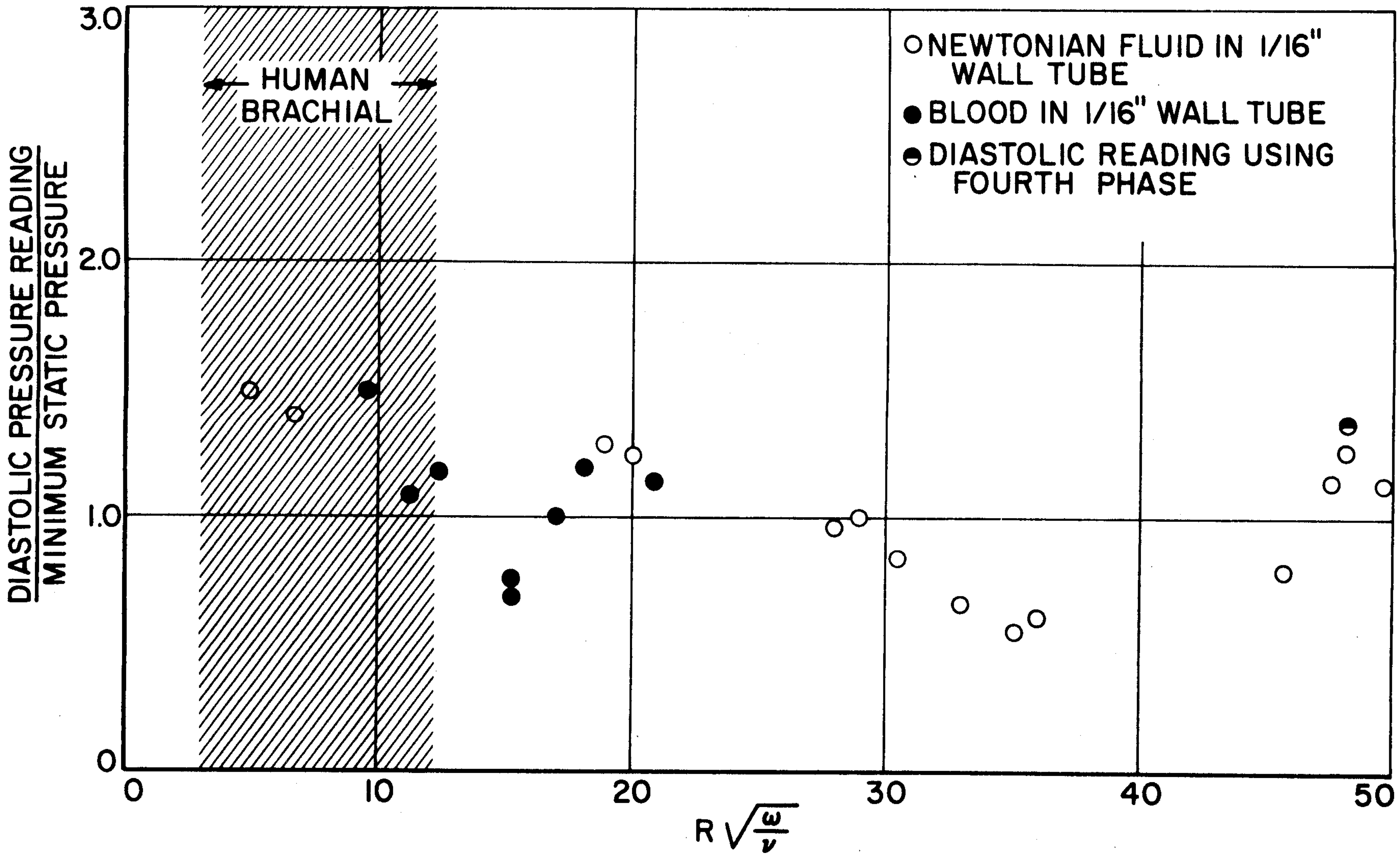

Figure 10.- Variation of diastolic pressure readings with frequency parameter for simulated arteries (mean pressure 155 mm Hg, pulse pressure 48 mm Hg).